IN VIVO RAPID SPECIFIC FORCE OF THE QUADRICEPS FEMORIS MUSCLE

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ABSTRACT

Luke R. Arieta: In vivo rapid specific force of the quadriceps femoris muscle
(Under the direction of Eric D. Ryan)

Maximal specific force measurements have shown utility but more recently the functional relevance of rapid contractions has been emphasized. Thus, the purpose of this study was to assess specific force at rapid time points and determine its reliability and relationship with sprint performance. Thirty-eight participants enrolled in the study and visited the laboratory on three occasions. They completed a sprint assessment, a dual energy X-ray absorptiometry scan, isometric strength testing of their thigh muscles, and ultrasound scans of their quadriceps. The reliability of the rapid specific force measurement increased with time, intraclass correlation coefficient, 0.443-0.679, standard error of measurement as percentage of the mean, 23.04%-63.19%. Sprint time was only related to absolute torque at 150 ms (r=-0.353, *P*=0.041) and 200 ms (r=-0.403, *P*=0.018). Future studies are needed to determine if rapid specific force is predictive of other performance metrics or if it can account for sex differences in rapid strength.

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LIST OF ABBREVIATIONS

ANOVA	Analysis of variance
BM	Body mass
BMI	Body mass index
CSA	Cross sectional area
DEXA	Dual energy x-ray absorptiometry
EMG	Electromyography
FL	Fascicle length
F_{PT}	Patellar tendon force
ICC	Intraclass correlation coefficient
LE	Leg extensors
LE_{RSF}	Rapid specific force of the leg extensors
LE TQ_{NET}	Net extension torque produced by the leg extensors
LF	Leg flexors
LF_{ACT}	Leg flexor muscle activation
LF_{AG}	Leg flexor torque during leg flexion
LF_{ANT}	Leg flexor torque during leg extension
$LF_{CO\text{-}ACT}$	Leg flexor co-activation
MA_{PT}	Patellar tendon moment arm length
MHC	Myosin heavy chain
MRI	Magnetic resonance imaging
MV	Muscle volume
MVC	Maximal voluntary contraction

M wave	Maximal muscle compound action potential
PA	Pennation angle
PCSA	Physiological cross-sectional area
$PCSA_{CORR}$	Corrected physiological cross-sectional area
RF	Rectus femoris
RF_{MV}	Muscle volume of the rectus femoris
RFD	Rate of force development
RMS	Root mean square
RTD	Rate of torque development
SD	Standard deviation
SEM	Standard error of measurement
TFCP	Tibiofemoral contact point
TQ_{50}	Torque-time curve 50 ms from contraction onset
TQ_{100}	Torque-time curve 100 ms from contraction onset
TQ_{150}	Torque-time curve 150 ms from contraction onset
TQ_{200}	Torque-time curve 200 ms from contraction onset
VI	Vastus intermedius
VI_{MV}	Muscle volume of the vastus intermedius
VL	Vastus lateralis
VL_{MV}	Muscle volume of the vastus lateralis
VM	Vastus medialis
VM_{MV}	Muscle volume of the vastus medialis

CHAPTER I: INTRODUCTION

Maximal voluntary muscular strength has traditionally been assessed to determine muscular performance (1). While maximal strength is important, many reviews and original research articles have suggested that rapid strength, the ability to increase force or torque as quickly as possible from rest, may be more related to sport-specific and functional tasks, more protective of common sport injuries, and more sensitive to detect acute and chronic changes in neuromuscular function (2-13). For example, the ability to produce force rapidly has been shown to be a better determinate of fall history than maximal force in elderly (9), as well as a better predictor of functional outcomes in clinical populations (6) and injury risk factors in first responders (8). Further, rapid force characteristics have athletic relevance as they have been shown to discriminate starters from non-starters in American football players (12), potentially due to its significant relationship with athletic tasks (4,11). Additionally, rapid strength is often examined at early (< 100 ms) and late (100-200 ms) time intervals from the onset of contraction which have been suggested to be governed by unique physiological characteristics (7,14,15).

Many factors contribute to the external force produced by a muscle, including the length of the tendon moment arm (16), neural activation (14), co-activation of the antagonistic muscles (17), and the architecture and size of the agonist muscles (18). It has been suggested that controlling for these extrinsic factors, a measure termed specific force, can provide a more accurate assessment of the intrinsic muscle properties in-vivo, including muscle fiber type, muscle fiber distribution, and lateral force transmission (19,20). For example, while controlling for the aforementioned properties, Sims and colleagues (2018) found that vastus lateralis specific

force was lower in adults with achondroplasia (16.7 ± 6.0 N • cm^{-2}) when compared to healthy controls (23.6 ± 6.4 N • cm^{-2}), suggesting the muscle itself is affected by this condition. Significantly lower gastrocnemius specific force has been reported in the elderly when compared to younger populations which suggests the age-related decline in specific force was primarily responsible for the age related loss in strength (21). Further, previous work (22) has demonstrated that chronic resistance training improved vastus lateralis specific force from 27 ± 6.3 N • cm^{-2} to 32.1 ± 7.4 N • cm^{-2} in older adults, along with similar results within younger adults (23), and the authors concluded that the change in specific force was responsible for the variable change in maximal strength. These studies among others, demonstrate the utility of maximal specific force assessments to determine the muscle's intrinsic force producing capacity in-vivo in clinical, young, and elderly populations, as well as following training routines. However, previous studies have assessed specific force during maximal contractions, and as previously stated, rapid contractions may be more functionally relevant. Thus, a reliable method of assessing rapid specific force may offer unique insight into inter-individual differences in the intrinsic muscle properties that govern the ability to rapidly produce force, which may be stronger predictors of functional performance (e.g. sprint time) than traditional measures of rapid force production as suggested by previous authors (24,25)

As noted in previous studies (19,20,22,26–30), maximal specific force assessments include measures of pennation angle and fascicle length that are measured during active contractions (during a MVC plateau). However, examining muscle architecture near the onset of contraction is challenging with traditional ultrasound devices and as such, muscle architecture near the onset of contraction will need to be determined from regression equations utilizing muscle architecture measures at maximal and various submaximal intensities. Thus, the

purposes of the current project were to: (1) create a regression equation to determine quadriceps muscle architecture during rapid force assessments, (2) determine the test-retest reliability of a novel rapid specific force calculation at commonly examined early and late time intervals, and (3) determine if rapid specific force variables explain more of the variance in sprint performance than traditional rapid force.

Research questions

1. Can rapid specific force at early and late time intervals be consistently examined?
2. Do rapid specific force variables explain more of the variance in sprint performance than traditional rapid force variables?

Research hypotheses

1. Rapid specific force at early and late time intervals will demonstrate acceptable absolute and relative consistency values.
2. Rapid specific force variables will be more related to sprint performance than traditional rapid force variables.

CHAPTER II: REVIEW OF LITERATURE

Specific Force

Specific force or tension refers to the force produced per unit area of muscle. This section will first briefly discuss in-vitro specific force and then, in-vivo specific force, which is more relevant to the current study. Traditionally, specific force has been assessed in-vitro from single, isolated muscle fibers where the peak tension is normalized to the fiber cross sectional area (CSA) (31–33). Usually, these experiments isolate and skin the fibers so they can be attached to a force transducer and artificially activated with saturating Ca^{2+} solutions to induce a maximal muscular contraction. A study by Luden and colleagues (2008) found that the specific force of isolated myosin heavy chain (MHC) I (109.0 ± 4.2 kN • m^{-2}) and MHC IIa (149.6 ± 6.7 kN• m^{-2}) fibers from the vastus lateralis muscle were significantly stronger than the MHC I (92.6 ± 4.4 kN • m^{-2}) and MHC IIa (130.4 ± 5.6 kN • m^{-2}) fibers from the soleus, respectively. Additionally, studies have displayed differences in the specific tension of different muscle fiber isoforms (34–36). For example, the specific tension of MHC I (43.77 ± 21.90 kN • m^{-2}) fibers from the vastus lateralis of adult males was found significantly lower than both MHC IIa (60.64 ± 34.86 kN • m^{-2}) and MHC IIb (61.84 ± 14.49 kN • m^{-2}). These studies assessing the specific force of isolated muscle fibers are in theory assessing the intrinsic force producing capacity of the fiber, as the only component contributing to the force produced is from the contractile machinery within the fiber. For example, it has been suggested that differences in single fiber specific force could be due to differences in the density of myofilament packing (37). While

these measurements are useful for examining differences in muscle fiber types and fibers from different muscles, the assessment of specific force in-vivo is more convoluted due to the numerous extrinsic factors that can influence the external joint moment produced.

Accurate in-vivo assessment of specific force requires that the external joint moment produced during a contraction be adjusted for the factors extrinsic to the muscle in order to assess the intrinsic force producing capacity of the muscle (38). For example, these factors include the tendon moment arm length (16), neural activation of the agonist muscle (14), co-activation of the antagonist muscle (17), muscle architecture and size (18). Further, an accurate assessment of in-vivo specific force requires that several of these factors, such as activation, co-activation, and architecture, be measured during an MVC in order to appropriately adjust the net joint moment produced. This was demonstrated in a previous study assessing the differences in specific force determined from both cadaveric and in-vivo assessments during MVC and reported that cadaveric measurements resulted in large overestimations of muscle force (force corrected for activation, co-activation, pennation angle, and moment arm length) and variable measurements of physiological cross sectional area (PCSA), which ultimately resulted in a significant over or underestimation in specific force of the tibialis anterior and soleus, respectively (26). Interestingly, specific force may decline through the developmental years, as one study has shown that boys had significantly higher specific force of the gastrocnemius muscle than men, 15.9 ± 2.7 N• cm^{-2} verses 13.1 ± 2.0 N• cm^{-2}, respectively. Although, these results are potentially muscle specific. A similar study found no difference in specific force of the quadriceps femoris muscle group between boys, girls, men or women (39). Regardless, specific force seems to be increased through training. Several studies have assessed the effects of resistance training on quadriceps femoris specific force (20,23,28,38). In addition to the 19%

increase in specific force displayed in older adults as previously discussed, young males increased specific force (17 ± 11%) of the quadriceps femoris from 25.9 ± 5.3 N• cm^{-2} to 30.3 ± 6.7 N• cm^{-2} following nine weeks of lower body resistance training (20). Resistance training has shown consistent results at increasing quadriceps femoris specific force, with the results ranging from 16.5-20.1% improvements (20,23,28,38). It has been suggested that the wide variations in strength coming from training interventions can largely be explained by discrepancies in the training adaptations to specific force (28). Further, while controlling for all extrinsic factors mentioned, studies assessing the specific force of the vastus lateralis (VL) or the whole quadriceps femoris muscle group ranged from 21.5 – 35.5 7 N• cm^{-2} (19,20,23,27,28,30,38,40). Results from these studies demonstrate the utility of specific force measured during MVC.

Components of Specific Force

Moment Arm Length

Muscular contraction exerts a force transmitted through a tendon in order to produce rotational movement of a body segment about a joint, often referred to as the axis of rotation or more specifically the instantaneous screw axis (41). The moment arm length is the perpendicular distance from this axis of rotation to the action line of the tendon, which is the patellar tendon for the leg extensors (16). Assessment of the patellar tendon moment arm length (MA_{PT}) is necessary to reduce the net joint moment in order to calculate patellar tendon force, since the torque produced about the axis of rotation of the knee is equal to the product of the patellar tendon force and the MA_{PT} (27). Several methods have been proposed to calculate the MA_{PT} including the geometric imaging method, the tendon excursion method, and the direct load measurement method (16). Of these methods, the two-dimensional geometric imaging method around the tibiofemoral contact point (TFCP) is commonly used. The benefit of using the TFCP,

while not an actual physical point within the knee joint, is that it is considered to be the point at which the knee joint compressive and shear forces are applied, and because of this the net compressive and shear torque is equal to zero, thus eliminating their contribution to the net joint moment and allowing for assessment of patellar tendon force (16). Both magnetic resonance imaging (MRI) (23,27,28,38), and dual-energy X-ray absorptiometry (DEXA) (19,30) have been utilized for assessment of the TFCP and subsequent MA_{PT}. However, the use of MRI normally requires that the leg being scanned remains fully extended due to spatial limitations within the bore of the magnet or the coil around the knee (38). Because the leg is usually flexed during MVC, the MA_{PT} must be adjusted to match the MA_{PT} at the same knee angle (27). Further, the MA_{PT} has been suggested to increase by 14% from rest to MVC and should be accounted for in order to accurately determine patellar tendon force (42).

Neural Activation

Voluntary muscular contractions are elicited via stimulation from the somatic nervous system and control of muscular force is dependent on both the number of motor units recruited and the rate (rate coding) at which they are activated (43). Additionally, the type (isometric, concentric, eccentric), speed, and duration of the contraction performed can all modulate the neural signaling. For example, rapid isometric contractions are likely more initially dependent on rate coding than recruitment in order to increase force production, as initial motor neuron discharge rates have been shown to reach upwards of 115 Hz during ballistic contractions from rest (44). In comparison, during gradual ramp-like contractions where the force is slowly increased, the motor units initially recruited exhibited discharge rates much lower (~5-10 Hz) and reached lower peak rates than those units with a higher recruitment threshold (45). Further, training ballistic muscle contractions can also alter the motor neuron discharge rate. Twelve

weeks of rapid contraction resistance training of the ankle dorsiflexors at 30-40% MVC significantly increased motor neuron discharge rate during the first three intervals at the onset of contraction, and interestingly the percentage of motor neurons displaying doublets, electromyography (EMG) spike intervals less than 5 ms apart, increased from 5.2% to 32.7% (46). Additionally, aging seems to decrease motor unit discharge rates. The first three interspike intervals were decreased in a group of older adults compared to younger adults and this was coupled with a 26% longer time to peak torque, 48% lower absolute, and 33% lower relative rate of force development during ballistic contractions of the ankle dorsiflexors (47). However, while assessing motor neuron discharge rates during sustained maximal contractions, discharge rates of 30-60 Hz were examined and motor neurons firing closer to 60 Hz tended to cease firing sooner (48).

Muscle activation is often assessed by the twitch interpolation technique and/or EMG amplitude normalized to the maximal muscle compound action potential (M wave) (49,50). The twitch interpolation technique utilizes a superimposed percutaneous stimulation of the femoral nerve during leg extensor MVC (51). The magnitude of this superimposed twitch can be utilized in an interpolated twitch ratio along with a resting twitch in order to calculate voluntary activation based on the negative linear relationship between voluntary and evoked twitch force (49). Normal ranges for quadriceps voluntary activation utilizing the twitch interpolation technique in healthy young adults have been reported to range on average from 89-96% during MVC, suggesting that there is a residual force producing capacity of the quadriceps even while attempting to maximally produce force (19,23,28,30). It has also be suggested that muscle activation can be assessed by normalizing the EMG signal to the M wave, with a potential advantage being the assessment of individual muscles, such as the vastus lateralis, as opposed to

force based assessments that measure the entire quadriceps group (50). Voluntary activation of the leg extensors has been shown to decrease with age during MVC from 91.9 ± 5.9 % in young to 86.4 ± 7.7 % in older (52); although others have found no difference with age (53). However, a recent meta-analysis concluded that there is a significant decrease in muscle activation of the leg extensors during MVC with increased age (54).

Co-activation

Co-activation of the antagonist muscle or muscle group during agonist contraction can result in a decrease of the external net joint moment produced, and adjusting for this co-contraction allows for a more accurate assessment of the force produced by the agonist muscle only (17). While assessing the co-activation of the hamstrings, it was found that there was a significant increase in the agonist moment when controlling for the antagonist moment verses not controlling for it during both eccentric and concentric contractions (17). However, some level of co-activation of the hamstrings during leg extension is required and potentially preferred in order to stabilize the knee joint and protect cartilage by distributing the articular surface pressure, which without could increase the chance of knee injury (55). Assessment of hamstrings activation during different contraction intensities has shown a linear relationship between EMG signal and torque (17,38). Therefore, in order to adjust for co-activation of the hamstrings while the quadriceps are acting as the agonist, previous authors have suggested using the ratio of the root mean square (RMS) EMG amplitude from the hamstrings during co-contraction while acting as the antagonist over the amplitude while acting as the agonist during maximal leg flexion to adjust the torque produced during maximal leg flexion (i.e. a linear relationship) (38). Previous studies have reported an increase in co-activation of the hamstrings during leg extension with aging in both older men (56) and women (57). Interestingly, studies

assessing the effect of training on co-activation have yielded differing results. Some studies have found no difference in co-activation following resistance training in both younger (20,23,28) and older adults (38,58,59), while some have shown a decrease (60), or even an increase (61). Although, it should be noted that this increase was seen at the ankle joint, potentially suggesting a joint specific response to resistance training.

Muscle Architecture

The architecture of a muscle, including both pennation angle (PA) and fascicle length (FL), have both been shown to influence the force producing capacity of the muscle (62,63). Pennation angle is the angle at which the muscle fibers insert into the aponeurosis, and as the angle increases the force transmitted to the tendon is decreased (62,64). Correcting for the angle of pennation by utilizing the cosine of the PA allows for assessment of the true force produced by the fibers (40). Fascicle length has been reported to be a measure of the number of sarcomeres in series, and as such is representative of the maximal shortening velocity of the muscle (65). Pennation angle and FL have also been shown to change in a curvilinear manner with increasing isometric force production suggesting their contribution to muscular force output may change throughout a contraction, with a 23.5 ± 3.3 % decrease and 39.7 ± 6.6 % increase in FL and PA from rest to MVC reported, respectively (18). Further, trained individuals have been reported to have greater PA than untrained individuals (64). Studies that have implemented resistance training programs have shown an increase in PA pre to post intervention of the VL in both younger (20) and older adults (38); however, when all four muscle of the quadriceps group are examined, no difference was reported for the RF, VI, and VM in the young (23). Fascicle length of the VL was also shown to increase following resistance training in older adults (38). Aging alone has shown mixed results, with some studies reporting a decrease in both PA and FL

(66), while others report no change in either (67). Regardless, assessment of muscle architecture measures at the time of force being produced is required for accurate assessment of the specific force of a muscle (19,20,22,26–30).

Muscle Size

A primary factor responsible for the force producing capacity of a muscle is the number of sarcomeres is parallel (68). The anatomical CSA of non-pennate muscles will give an appropriate measure of the number of sarcomeres in parallel as the fibers would be intersected at right angles to the line of force, but this is an inaccurate measure in pennate muscles due to the angle at which the fibers insert into the aponeurosis (40). When taking into account the angle of the muscle fibers, a cross section taken so there remains a right angle between the fibers and the force they produce is termed the PCSA, which can be approximated by dividing the muscle volume (MV) by the FL (26,69,70). Multiple assessments of muscle imaging have been reported including the use of both MRI (71,72) and ultrasonography (38). Muscle volume is then calculated by utilizing the separate CSA images in an estimation equation. Although the number of images needed varies based upon the equation used, some of which utilize a single CSA scan (72), whereas others require several (71,73). However, other methods exist such as diffusion tensor imaging (74), to determine PCSA. Some studies have displayed an increase in PCSA of the quadriceps following training in younger adults (20,28); however, these studies utilized architecture of the VL only, and a study assessing all constituent muscles of the quadriceps found that some (RF and VM), but not all (VL and VI) increased PCSA following resistance training (23). Although, while accounting for the volume and architecture of each separate muscle independently, an overall increase of the quadriceps muscle group was shown following resistance training (23). However, previous work in older adults have reported no changes in VL

PCSA following resistance training in older adults due to the slightly greater increase in FL than MV (38).

Intrinsic Factors

Inter-individual differences in the specific force of the quadriceps muscle group, along with resistance training mediated increases, both suggest that the intrinsic force producing capacity of the muscles can differ between individual and be enhanced with training. However, exactly which intrinsic factors, such as muscle fiber type and contractile properties or lateral force transmission, are responsible for the differences and increases is not yet fully understood (20). The Huxley muscle model suggests that thick (myosin) and thin (actin) filaments slide past each other, thus changing the length of the sarcomere and subsequently the muscle, producing force (75). While this model explains how muscular force is produced, differences in the distribution of muscle fiber types may influence how much force in produced. Generally, there are two commonly utilized methods to determine muscle fiber type, which are by myosin heavy chain content or myosin ATPase (76). Commonly reported fiber types include both slow (type I) and fast (type IIa and IIx) isoforms, but several hybrid isoforms also exist, and the proportion of these hybrid fibers has been shown to increase with advancing age as motor unit remodeling, brought about by the denervation-reinnervation process, accelerates (77,78). Several studies have shown a higher specific force of isolated fast twitch fibers compared to the slow twitch fibers (31,34–36). Further, a study assessing the effects of one year of resistance training showed a 64% and 37% increase in the specific force of both type I and IIa fibers from the VL, respectively; however, this study reported no change in the CSA within each of the fiber types, and a decrease from 49% to 31% in the number of type IIa pre to post training (36). Similar results were found in older women after one year of resistance training as the specific force of

type I and type IIa fibers increased 24% and 37%, respectively (33). Nearly complete opposite results were found in a study assessing the adaptations to a shorter 12-week resistance training program. Results from this study displayed no difference in specific force within fiber types, an increase in the CSA of type I, IIa, and IIa/IIx, of which the absolute increase in type II fibers was greater but the relative increase was similar across fiber types, along with an increase in the percentage of type IIa fibers from 30% to 55% pre to post training (31). The length, one-year verses 12-weeks, could potentially explain the discrepancy seen in the results. A study by Erskine and colleagues (2011) assessing in-vivo specific force prior to and following resistance training found that along with an increase in whole muscle specific force, the proportion of myosin heavy chain IIx decreased from 29% to 19% with no change in any of the other isoforms; however, no correlation was found between in-vivo specific force and any of the myosin heavy chain isoform (i.e. I, IIa, or IIx) content or the change in MHC type IIx content (20). Interestingly, while the in-vivo specific force of the quadriceps did increase, the specific force of the isolated fibers across all fiber types did not change with training. The results from this study suggest that neither fiber type composition nor myofibrillar packing density, as assessed by the lack of change in the specific force of isolated fibers, are responsible for the increased in-vivo specific force. This study also found no change in the fiber peak power normalized to fiber volume, which alongside the increase in in-vivo specific force potentially suggests an increase in lateral force transmission, although this is an indirect measure.

Rapid Strength

Rapid or explosive strength refers to the amount of force/ torque that can be produced in the first 200 – 250 ms after the onset of contraction (7). Several measures of performance can be obtained from rapid contractions, including absolute and normalized rapid strength, rate of

force/torque development (RFD/ RTD) (i.e. slope of the force/torque-time curve), and impulse (11,79). Rapid strength of the leg extensors is usually assessed during isometric contractions with an isokinetic dynamometer utilizing a rotational torque transducer, although it has been suggested that custom built dynamometers utilizing linear strain gauge load cells may be more accurate due to the decreased compliance allowing for minimal changes in joint angle (7,80). However, it should be noted that custom built dynamometers may restrict the joint angle that is able to be tested, and thus may lack the ability to assess rapid strength at optimal joint angles. Previous studies have shown differences in both absolute and normalized rapid strength and maximal strength variables based on knee joint angle (81–83).

Physiological Determinants

Rapid strength assessments are usually measured at early (i.e. <100 ms) and late (100-200 ms) phases, and are said to be governed by unique physiological parameters (7,14,15). For example, a study by Andersen and Aagaard (2006) assessed the relationship between RFD and various physiological parameters, such as MVC and evoked twitch RFD. As the time from the onset of contraction increased, the relationship between MVC and voluntary RFD increased and MVC explained greater than 50% of the difference in voluntary RFD at time points greater than 90 ms. Additionally, twitch RFD and voluntary RFD displayed a positive relationship at time points up to 50 ms, but was non-significant as time increased. These data suggest that late phase rapid force may be more reliant on MVC, whereas early phase may be more reliant on intrinsic muscle properties. However, it is important to consider that while evoked twitches eliminate voluntary neural drive to the muscle, other factors, such as muscle architecture and size could influence the net torque produced, and thus not exclusively measure the intrinsic muscle properties (18). Although, a previous study has demonstrated a significant relationship between

evoked twitch RFD and intrinsic contractile properties of the muscle assessed by MHC isoforms, but this relationship was only shown while assessing this relationship across multiple muscle groups (84). Regardless, evoked twitch RFD only explained 36% ($r^2 = 0.36$) of the variance in voluntary RFD during the early phase, suggesting other factors are likely also responsible (15). A potential explanation is that neural activation is also responsible for the variance in rapid strength variables during the early phase of contraction. A study assessing the association between rapid strength and muscle activation, assessed by RMS EMG normalized to the maximal M wave, found that the early phase of explosive contractions was positively associated with muscle activation (80). The same study also reported an increasing reliance on MVC as a rapid contraction progressed, along with an evoked octet being the primary determinant of the RFD from 50-100 ms. Further, motor unit modeling has suggested that the motor unit discharge rate may be the primary determinant of muscle activation responsible for augmented RFD (85). These results further suggest that the early phase of rapid contractions may be reliant on a number of factors, such as motor unit discharge rate and intrinsic muscle properties, while the later phase is primarily reliant on MVC. Additionally, stiffness of the tendon aponeurosis complex may also be a determinant of rapid strength, as a study has shown a positive relationship with RTD during 0-100 ms ($r = 0.54$) and 0-200 ms ($r = 0.56$), or the late phase of a rapid contraction (86).

Performance and Injury Prevention

Several studies have assessed the role of rapid strength on athletic performance (4,5,10–13). For example, a study assessing rapid strength variables of the leg extensors and flexors found that only rapid strength characteristics of the leg flexors, specifically the time to peak RTD, RTD at 30 ms, impulse at 30 and 50 ms, and absolute torque at 30 ms, could discriminate

starters from non-starters among collegiate American football players (12). Similarly, a study assessing RTD of the hip extensors was able to discriminate female collegiate soccer players and non-athlete controls based on absolute and normalized RTD from 0-50 ms, but not RTD at 100-200 ms (10). A study assessing explosive force production during isometric squats in rugby union players found that absolute and normalized rapid force at 100 ms (i.e. early phase) was significantly related to both 5 m (absolute; normalized: r = -0.50; r = -0.63) and 20 m (absolute; normalized: r = -0.50; r = -0.54) sprint (4). Rapid strength may also have functional implications for elderly individuals. Older adults with a higher rate of force development of the plantar flexors have been shown to have higher maximal walking speeds (2). Also, older women without a history of falls have been shown to have a higher absolute (P = 0.039) and relative (P = 0.011) RTD of the hip extensors at 0-50 ms than those with a history of falls (9). Interestingly, RTD at 100-200 ms was not predictive of fall history in this group of older women, potentially suggesting the importance of early phase rapid strength in preventing falls, specifically in older adults. Additionally, rapid strength has been utilized to assess injury risk factors in first responders. Mota and colleagues (2018) found that both absolute and normalized later phase (100-200 ms) rapid strength of the quadriceps was related (r = -0.462 - -0.336) to a performance index assessing functional balance performance in career firefighter. Results from this study highlight the relevance of rapid strength on occupational health and safety.

Training and Aging

Resistance or explosive training has been shown to be effective at increasing rapid strength variables (14,87). A study assessing the effects of 14 weeks of resistance training found that RFD increased at 30, 50, 100, and 200 ms, while normalized RFD increased 15% from onset to one-sixth MVC along with a decrease in the time to one-sixth MVC (14). Although the type

of training completed will influence the adaptive response. For example, maximal strength training has been shown to increase MVC to a greater extent than explosive strength training, but the opposite was true for explosive force, particularly at early time points (i.e. 100 ms) (87). While training may augment rapid strength, aging has been shown to have a deleterious effect on rapid strength. Older men (66.8 ± 4.5 years) have been shown to have decreased RTD of the leg extensors at 50, 100, and 200 ms along with diminished peak RTD compared to both young (24.9 ± 3.0 years) and middle-aged (50.6 ± 4.0 years) men (88). However, no age-related differences were shown in relative rapid strength variables. A potential explanation for this is the decreased peak torque in the older compared to both the middle-aged and younger group, which is likely related to the lower estimated total thigh muscle CSA. Older men have also displayed lower absolute and relative plantar flexor RTD at 100-200 ms but not 0-50 ms compared to young (79). Also, this lower absolute RTD at 100-200 ms was related to higher echo intensity from ultrasonography, smaller PA, and a lower RMS EMG amplitude. Relative RTD at 100-200 ms was also related to echo intensity and RMS EMG amplitude, but not PA. A potential explanation for the lack of difference between the young and older group for RTD at 0-50 ms is that muscle activation, assessed by RMS EMG, was not different between groups, suggesting that while similar at contraction onset, older individuals may have a diminished ability to sustain motor unit discharge rates after the onset of contraction.

Test-retest reliability

Test-retest reliability refers to the consistency of a test in measuring the same variable across multiple testing sessions, usually across multiple days (89). However, there are two types of consistency that must be considered. Relative consistency refers to the rank of an individual's scores in relation to other individuals being tested, which is quantified by the intraclass

correlation coefficient (ICC) (89). An ICC of 1.0 would indicate that perfect reliability exists, whereas an ICC of 0 would indicate that the test is not reliable at all. Because the calculation of ICC is influenced by between-subjects variability, if there is little differences between the scores of all participants then the ICC is likely to be small, and the opposite is true for samples with large differences (89). Thus, ICC is best used as a measure of test-retest reliability in combination with a measure of absolute consistency, such as the standard error of measurement (SEM). Absolute consistency refers to the consistency of the results across multiple test of one individual and can be used as a measure of the precision of a score (89). The SEM should be calculated as the square root of the mean square error term from an analysis of variance (ANOVA) in order to be independent of the ICC (90). There are numerous models available to calculate both ICC and SEM, but model "2,1" from Shrout and Fleiss (91) is often utilized in the assessment of test-retest reliability data for physiological performance measures in which the results are to be generalized to other laboratories. Using model "2,1" assumes that the data contains a single score from each trial for each participant and that all subjects were assessed by the same group of raters, whom were randomly selected from the population of raters. This allows for results of the test-retest reliability to be generalized to other groups of raters or laboratories. Furthermore, authors also suggest using a one-way within-subjects ANOVA to assess systemic error (i.e. unidirectional change in performance on repeated testing, such as the learning effect or fatigue) (89). If systemic error is found to exist, additional testing visits may need to be included until there is a plateau in performance in order to eliminate such bias (89).

CHAPTER III: METHODOLOGY

Participants

Thirty-eight young (21.8 ± 3.1 years) non-obese (BMI: 23.3 ± 2.4 kg/m^2; stature: 172.4 ± 10.2 cm; body mass: 69.8 ± 12.8 kg) men (n = 16) and women (n = 22) from the University of North Carolina at Chapel Hill surrounding areas participated in the current project. Participants were recreationally active (defined as more than one but less than five hours per week of exercise), free of any neuromuscular, cardiovascular, or metabolic disease and had no history of major orthopedic surgery, previous knee injury, or serious pain in the lower extremities while performing lower extremity exercise. Female participants were eumenorrheic for the previous six months leading up to enrollment and not pregnant nor trying to become pregnant. This study was approved by the University of North Carolina at Chapel Hill Institutional Review Board (#18-2137) and all participants read and signed an informed consent document prior to all study activities.

Experimental Design

Participants visited the laboratory on three occasions for a familiarization (visit one) and two subsequent testing sessions (Figure 1). There was 2-10 days between visit one and the first testing session (visit two), and 3-10 days between visit two and the second testing session (visit three) to account for potential residual muscle soreness. Upon arrival for visit one, all participants 1) read and signed an informed consent document, 2) completed a health history questionnaire, 3) had their stature and body mass measured, 4) were familiarized with the

maximal strength testing protocol and submaximal torque tracing, and 5) completed a 10-meter sprint assessment. Participants arrived to visit two and three having abstained from strenuous exercise (48 hours), alcohol (24 hours), caffeine (12 hours), as well as fasted (four hours) prior to testing. Testing for visit two and three included 1) a dual-energy x-ray absorptiometry (DEXA) scan of the knee, 2) ultrasonography of the participant's thigh at rest and, 3) maximal isometric strength testing of the leg flexors and extensors. In addition, at the end of visit two, participant's completed submaximal torque tracings while simultaneously using ultrasonography to examine muscle architecture of the superficial muscles of the thigh.

Stature and Body Mass

Stature was measured to the nearest 0.5 cm and body mass (BM) was recorded to the nearest 0.1 kg using a calibrated stadiometer and clinical scale (Seca 769, Seca, Chino, CA), respectively. Body mass index was calculated according to the following equation: BMI = body mass (kg) / height^2 (m^2).

Isometric Strength Testing

Participants were seated on a calibrated isokinetic dynamometer (HUMAC Norm, Computer Sports Medicine Inc., Stoughton, MA) with a 115° angle between their torso and thigh. Restraining straps were placed and tightened across the torso and pelvis to minimize participant movement during contractions. Participants were asked to cross their arms in front of their chest and refrain from grabbing the handles at their sides to provide leverage during contractions. The dominant leg was extended to 60° below the horizontal plane and secured to the lever arm five cm above the lateral malleolus of the ankle. The lateral epicondyle of the femur was aligned with the axis of rotation of the dynamometer. Participants completed three warmup contractions (50 – 75% of perceived maximum strength) of the leg extensors (LE)

followed by two MVC's held for 3-4 seconds with two minutes of rest between each contraction. Participants were asked to "kick as hard and fast as possible" and were verbally encouraged throughout. Torque (Nm) signals were sampled at 2,000 Hz with a Biopac acquisition system (MP150, Biopac Systems, Inc., Santa Barbara, CA) and stored offline on a personal computer for processing with a custom written program (LabVIEW 2018, National Instruments, Austin, TX). Torque was gravity corrected for passive limb weight and filtered using a fourth order, zero phase shift low pass Butterworth filter with a 150 Hz cutoff (92). Each MVC was visually inspected for countermovement or pretension (8,79). The average baseline torque slope of the 200 ms preceding contraction onset was -0.27 ± 2.47 Nm •s^{-1}. Contraction onset was determined manually using a high-resolution x- and y-axis scale as described previously (7). A vertical cursor was placed at the point where the signal deflected (i.e. last trough before signal deflection) from baseline (7). Rapid torque variables were calculated from the contraction with the highest peak TQ value using the torque-time curve at 50 ms (TQ_{50}), 100 ms (TQ_{100}), 150 ms (TQ_{150}), and 200 ms (TQ_{200}) from onset (8,79). These timepoints were used because they represent unique physiological characteristics that govern rapid torque development. Early rapid torque time points (< 100ms) are primarily determined by neural contributions, such as motor unit discharge rate, and the later time points (>100 ms) are more influenced by MVC force (7).

Coactivation

Surface EMG of the biceps femoris was used to represent the entire leg flexor (LF) muscle group and was measured with a surface bipolar, pre-amplified electrode (TSD150B Biopac Systems, Santa Barbara, CA; gain = 350 and interelectrode distance of 20 mm). The electrode was placed one-half the distance from the ischial tuberosity to the lateral epicondyle of the tibia, in accordance with guidelines from the Surface EMG for the Non-invasive Assessment

of Muscles project (93). A pre-gelled disposable reference electrode (Ag-Ag Cl, Quinton Quick Prep, Quinton Instruments Co., Bothell, WA) was placed on the tibial tuberosity to serve as a ground electrode. Prior to electrode placement, the skin was shaved, lightly abraded, and cleaned with isopropyl alcohol wipes and secured to the skin with hypoallergenic tape. Leg flexor EMG was recorded during three LF MVCs utilizing the same protocol as described above (see Isometric Strength Testing) and during all LE MVCs to measure LF coactivation. All LF MVCs were completed with the leg extended to 60° below the horizontal plane in order to be at the same angle as during LE contractions (17). All EMG signals were sampled at 2,000 Hz and filtered with a fourth-order, zero phase shift Butterworth filter with a band pass of 10-500 Hz. Leg flexor peak torque was determined from the highest 250 ms epoch during three, 3-4 second LF MVCs, and the contraction with the highest peak torque was selected for further analysis. Rapid torque variables were calculated from the LF (LF_{AG}) at the same timepoints as the LE (50 ms, 100 ms, 150 ms, and 200ms). Leg flexor activation (LF_{ACT}) and coactivation ($LF_{CO\text{-}ACT}$) were calculated as the RMS of the EMG signal during 50 ms intervals centered around each LF and LE rapid torque time point, respectively (80). The exact intervals were 25-75 ms, 75-125 ms, 125-175 ms, and 175-225 ms from EMG onset for TQ_{50}, TQ_{100}, TQ_{150}, and TQ_{200}, respectively. The onset of EMG signal was determined as the last trough or apex prior to signal deflection, independent of torque onset (79).

Leg flexor torque (LF_{ANT}) during LE MVC at each rapid torque timepoint was calculated based upon a linear assumption between torque and EMG of the LF according to the following equation (17,22): $LF_{ANT} = (LF_{CO\text{-}ACT} / LF_{ACT}) \times LF_{AG}$

Muscle Volume

A portable, brightness mode ultrasonography device (LOGIQ e 5, General Electric Company, Milwaukee, WI) with a multi-frequency linear array probe (12L-RS, 5-13 MHz, 38.4 mm field of view, General Electric Company, Milwaukee, WI) was used to measure muscle CSA (cm^2) of the RF, VL, VI, and VM. Participants lay supine on an exam table while CSA scans were taken at rest with minimal pressure applied and acoustic coupling gel to increase image quality. Panoramic images were taken at 25%, 50%, and 75% of muscle length for the RF, VL, and VI; 33%, 50%, and 66% of muscle length were used for the VM. Different percentages of muscle length were used for the VM due to pilot testing that revealed the VM has a different shape than the other three quadriceps muscles (RF, VL, and VI), which made distal scans less visible. Images were analyzed in an open source image program (ImageJ, National Institutes of Health, Bethesda, MD) and the straight-line function was used to convert pixels to cm. Each muscle (RF, VL, VI, and VM) was outlined with the polygon function to include the most amount of muscle tissue as possible and least amount of surrounding fascia, which was then analyzed to determine CSA. Muscle volume (cm^3) was calculated for each of the individual quadriceps muscles (RF_{MV}, VL_{MV}, VI_{MV}, and VM_{MV}) using the Cavalieri formula shown below, which assumes a cylinder shape of the muscle between slices (73).

$$MV = \sum_n e_i \times CSA_i$$

Where n is the number of slices used, and e_i is the distance between available slices i and i + one.

Muscle Architecture

Muscle architecture was assessed using the same ultrasonography device and isokinetic dynamometer set up as described previously (see Muscle Volume and Isometric Strength

Testing, respectively). Ultrasound scans were taken while the participant was seated on the isokinetic dynamometer at 0% (rest) and at 20%, 40%, 60%, 80%, and 100% MVC. A line representing the appropriate percentage MVC was displayed on a monitor in front of the participant (for 20%, 40%, 60%, and 80% only) and they were asked to trace this line with the live digitized torque signal. Once the participant reached the desired torque level and was steadily tracing the line, two investigators simultaneously took panoramic scans along the fascicular plane following the muscle trajectory proximal to distal of the VL/VI and VM. An additional scan was performed during an additional torque tracing to capture the RF. The scans at rest and at 100% MVC were completed first and second, respectively, with the remaining scans completed in random order to minimize the order effect. Images were analyzed similar to the methods described previously (see Muscle Volume). Fascicle length (cm) was measured as the length of the fascicle from the superficial to the deep aponeuroses just inside the surrounding fascia (79). Two fascicles that could clearly be seen from the middle of each image were measured and averaged. Using the same two fascicles, pennation angle (°) was determined as the angle between the fascicle and the deep aponeurosis and the average of the two was determined for further analysis (79,94). Since the RF is a bipennate muscle, FL and PA were determined from the lateral side only with the assumption that the medial side had identical architecture (27). Due to the curvilinear relationships between FL and PA with torque, second-order polynomial regression equations were established for both FL (Figure 2) and PA (Figure 3) of each muscle (VL, VI, RF and VM) using the data points from rest, 20%, 40%, 60%, 80%, and 100% MVC (95). Fascicle length and PA at each rapid time point (50 ms, 100 ms, 150 ms, and 200 ms) were then estimated from the second-order polynomial curves based on the torque produced at each rapid time point relative to maximal torque.

Physiological Cross-Sectional Area

The PCSA (cm^2), which is representative of the number of sarcomeres in parallel, of each of the four quadriceps muscles (RF, VL, VI, and VM) was calculated at each of the four rapid time points utilizing the FL specific to each based on the following equation (27,96): PCSA = MV / FL

Muscle volume of the RF was divided by the lateral FL (see Muscle Architecture) (27).

Patellar Tendon Moment Arm Length

Patellar tendon moment arm length (MA_{PT}; cm) was measured with a daily calibrated DEXA (Lunar iDEXA, GE Healthcare, Chicago, IL) as described previously (30). Participants lay on the exam table on the side of their dominant leg with their knee flexed to a 60-degree angle using a goniometer (Baseline Plastic Goniometer 12-1000, Fabrication Enterprises Inc., White Plains, NY) to match the angle used during isometric strength testing. Images were exported to an open source image program (ImageJ, National Institutes of Health, Bethesda, MD) and the straight-line function was used to convert pixels to cm. Moment arm length was measured as the perpendicular distance between the tibiofemoral contact point, the midpoint of the distance between the two femoral condyles and the tibial plateau, and the patellar tendon action line (16).

Calculation of Rapid Specific Force

Specific force was calculated as previously described (27,30). Rapid specific force was calculated at each rapid time point (50 ms, 100 ms, 150 ms, and 200 ms) using the TQ, LF_{ANT}, PCSA, and PA data specific to the rapid time point being used. Torque produced by the LE at each rapid time point was corrected for LF torque at the same time point to determine a net torque value using the following equation: LE TQ_{NET} = TQ + LF_{ANT}

Where LE TQ_{NET} is the net extension torque produced by the LE.

Patellar tendon force (F_{PT}) was calculated using the following equation: F_{PT} = LE TQ_{NET} / MA_{PT}

The PCSA of each quadriceps muscle was corrected for its PA specific to each rapid time point, as this reduces the amount of force transmitted to the patellar tendon, and summed to give a total corrected PCSA ($PCSA_{CORR}$) based on the following equation: $PCSA_{CORR} = \sum$ PCSA x cosine (PA)

Finally, rapid specific force of the LE (LE_{RSF}) at each time point was calculated using the following equation: $LE_{RSF} = F_{PT} / PCSA_{CORR}$

Sprint Assessment

Thirty-four participants performed a maximal sprint assessment. Beginning at the start line, participants started on their own action from a standing position and sprinted as fast as possible 10 meters to the finish line. Ten-meter sprint time (sec) was assessed from timing gates (SMARTSPEED PRO, Fusion Sport, Boulder, CO) that were placed at the start and finish line. Prior to the sprint assessment, participants completed a warm-up and one practice trial prior to three maximal sprints with one minute of rest between each. The fastest time of the three sprints was used as 10-meter sprint time.

Statistical Analysis

Test-retest reliability for LE_{RSF} at each time point was determined from days two and three per the methods described by Weir (89). One-way repeated measures ANOVA were used to determine systematic variance across testing visits. Model "2,1" from Shrout and Fleiss (91) was used to examine relative and absolute consistency with the ICC and SEM, respectively. The SEM was calculated as the square root of the error mean square from the ANOVA in order to be

independent of the ICC (90). The SEM values were also expressed as a percentage of the mean. Pearson's product moment correlation coefficients were used to assess the relationship between LE_{RSF} and traditional rapid force variables for each of the four timepoints (50, 100, 150, and 200 ms) from visit two with 10-meter sprint time. These relationships were also assessed with the LE_{RSF} and traditional rapid force variables normalized to body mass, per recent recommendations (11). If both traditional and rapid specific force variables were significantly related, a Steiger's Z was used to assess if LE_{RSF} was more related to the sprint time than traditional rapid force variables (97). All analyses were performed with SPSS version 26.0 (IBM SPSS Inc., Chicago, IL, USA) with statistical significance determined *a priori* at an alpha level of $P \leq 0.05$.

CHAPTER IV: RESULTS

Three participants withdrew from the study following visit two, two data sets were dropped from visit three due to excessive pre-onset slopes, and we were unable to capture sprint times from four participants, muscle architectural measures of the VL for three participants, VI for six participants, and RF for four participants. Thus, data from 38, 35, 34, and 32 participants were used to create the regression equations for FL and PA of the VM, VL, RF, and VI, respectively, 34 participants were used to assess the relationships with sprint time, and 33 participants were used to assess the test-retest reliability. All data is presented as mean ± standard deviation. All LE_{RSF} and absolute rapid torque values used to assess test-retest reliability across testing days are shown in Table 1. Systematic variance was not present across testing days for either the LE_{RSF} at any time point ($P \geq 0.282$) or for absolute rapid torque at any time point ($P \geq 0.429$). The ICC for LE_{RSF} was 0.443, 0.570, 0.563, and 0.679 for 50 ms, 100 ms, 150 ms, and 200 ms, respectively. All SEM data is presented as SEM (% of mean). The SEM for LE_{RSF} was 0.68 N•cm^{-2} (63.19 %), 2.96 N•cm^{-2} (49.70 %), 3.30 N•cm^{-2} (35.51 %), and 2.70 N•cm^{-2} (23.04 %) for 50 ms, 100 ms, 150 ms, and 200 ms, respectively. The ICC for absolute rapid torque was 0.462, 0.433, 0.710, and 0.759 for 50 ms, 100 ms, 150 ms, and 200 ms, respectively. The SEM for absolute rapid torque was 2.26 N•m (45.33 %), 24.42 N•m (63.28 %), 20.66 N•m (33.66 %), and 19.97 N•m (24.81 %) for 50 ms, 100 ms, 150 ms, and 200 ms, respectively. All LE_{RSF}, absolute rapid torque, and their normalized counterparts from visit two used to assess the relationship with sprint time (2.03 ± 0.22 secs) are presented in Table 2. Lower sprint time (better performance) was negatively related to higher absolute rapid torque at

150 ms (r = -0.353, $P = 0.041$) and 200 ms (r = -0.403, $P = 0.018$) (Figure 4), but not at 50 or 100 ms ($P \geq 0.176$). Sprint time was not related to LE_{RSF} ($P \geq 0.429$), body mass normalized LE_{RSF} ($P \geq 0.202$), or body mass normalized absolute rapid torque ($P \geq 0.109$) at any time point.

CHAPTER V: DISCUSSION

The purpose of the current study was to extend the work of previous authors (20,23,27–30,38,39,96) who have examined maximal specific force and create a model to assess rapid specific force in young healthy men and women. One key objective of the current study was to create regression equations that could be used to determine muscle architecture at time intervals near contraction onset commonly used to examine rapid muscle strength. Eight regressions equations were created to assess FL (Figure 2) and PA (Figure 3) at rapid time points for the VL, VI, RF, and VM, based on architectural measures of each muscle at rest, 20 %, 40 %, 60 %, 80 %, and 100 % of MVC. The data were fit with second-order polynomial regressions equations (R^2 = 0.9733-0.9987). Similar to the results from a previous study, as torque increased from rest to MVC, FL decreased and PA increased for all four muscles of the quadriceps in a curvilinear manner (95). Although, the FL decreased slightly more from rest to MVC, -29.2 %, -37.6 %, -39.5%, and -24.2 %, for the VL, VI, RF, and VM, respectively, in the current study than previously reported of -24.1%, -24.7 %, -20.6 %, and -21.6 % for the same muscles, respectively. Similarly, the relative increase in PA from the current study of 52.6 %, 55.5 %, 75.6%, and 67.5% from rest to MVC for the VL, VI, RF, and VM, respectively, was also slightly higher than previously reported of 24.1 %, 47.1 %, 58.6 %, and 45.1 % for the same muscles (95). While our data follow the general trend from the previous study, two key methodological differences may be driving the discrepancy. First, the current study utilized an extended field of view and panoramic mode ultrasonography, which allowed for the entire FL to be captured in one image. This is in contrast to the previous study which utilized a linear extrapolation technique to

calculate total FL by summing the portion visible in the image and the portion that extended out of the field of view (95). This technique of linear extrapolation has previously been shown to consistently underestimate true FL (98). Second, the current study assessed the distal portion of the VM, the vastus medialis oblique, which has been shown to have a significantly greater PA than the proximal VM, the vastus medialis longus (99). Nonetheless, these equations allowed for the calculation of the novel LE_{RSF} measurement and the assessment of its reliability.

Our findings indicated that there was no systematic error between testing days for either the LE_{RSF} ($P \geq 0.282$) or absolute rapid torque ($P \geq 0.429$) values. The ICC values for LE_{RSF} were 0.443, 0.570, 0.563, and 0.679 for 50 ms, 100 ms, 150 ms, and 200 ms, respectively. However, the ICC alone is only representative of the relative consistency across testing days and as suggested by Weir (89), should be evaluated in conjunction with the SEM. The SEM values for LE_{RSF} were 0.68 $N \cdot cm^{-2}$ (63.19 %), 2.96 $N \cdot cm^{-2}$ (49.70 %), 3.30 $N \cdot cm^{-2}$ (35.51 %), and 2.70 $N \cdot cm^{-2}$ (23.04 %) for 50 ms, 100 ms, 150 ms, and 200 ms, respectively. When taken together, overall reliability of the LE_{RSF} measure improves with increasing time from the onset of contraction. Due to the novelty of the LE_{RSF} measurement, assessing reliability of the results from the current study to previous studies is difficult. One previous study that assessed specific force during a MVC reported an ICC value of 0.74 and coefficient of variation of 8.79% (27), which is lower than many previous studies examining traditional MVC isometric strength, ICC 0.95-0.99 (27,38,100). Thus, it is possible that specific force measurements are less reliable than traditional strength measurements. Additionally, rapid force variables may be inherently less reliable than maximal strength measurements, which was suggested in a recent review that noted rapid strength reliability was commonly lower when compared to MVC assessments, especially at early time points (7). The absolute rapid torque from the current study had ICC values of

0.462, 0.433, 0.710, and 0.759, along with SEM values of 2.26 N•m (45.33 %), 24.42 N•m (63.28 %), 20.66 N•m (33.66 %), and 19.97 N•m (24.81 %) for 50 ms, 100 ms, 150 ms, and 200 ms, respectively, which are all poorer than the ICC and SEM for MVC of 0.913 and 16.39 Nm (9.86%), which is in agreement with the review (7). Several previous studies have also assessed the reliability of rapid contraction variables and reported absolute and relative consistency values lower than that of maximal strength (4,100,101). For example, a study assessing explosive isometric squat reported ICCs and (coefficient of variation) of 0.74 (14.6 %), 0.96 (7.4 %), 0.88 (5.3 %), and 0.78 (7.9%) for 50 ms, 100 ms, 150 ms, and 200 ms, respectively (4). All of which were poorer than the ICC and (coefficient of variation) reported for MVC strength of 0.96 (4.0%). Another study that assessed unilateral leg extension force at 50 ms, 100 ms, and 150 ms reported ICC and (coefficient of variation) values of 0.80 (16.6%), 0.91 (6.4%), and 0.90 (5.1%), respectively, which all were poorer than MVC strength which was 0.95 (3.3%) (100). It is possible that the lower reliability of the rapid strength measurements is the use of a manual verses automated onset selection, which has previously demonstrated higher reliability (101). For example, an ICC and SEM (SEM%) of 0.652 and 8.83 N (29.69%), respectively, was reported for force at 50 ms in the plantar flexors utilizing manual onset compared to 0.862 and 3.17 N (11.77%), respectively, utilizing an automated onset detection (101). Additionally, the reliability of the absolute rapid torque values from the current study were considerably lower than those for both MA_{PT} (ICC, SEM (SEM %): 0.980, 0.06 cm (1.35%)) and MV of the quadriceps femoris (0.992, 37.54 cm^3 (2.74%)), further suggesting that the poorer reliability of the absolute rapid torque was mostly responsible for the reliability of the LE_{RSF} measurement. Taken together, the inherently lower reliability of both specific force and of rapid strength

measurements is likely responsible for the lower reliability of LE_{RSF} when compared to traditional maximal strength measurements.

The results from the correlations show that higher absolute rapid torque at 150 ms (r = -0.353, $P = 0.041$) and 200 ms (r = -0.403, $P = 0.018$) were related to lower sprint time, however, none of the other rapid torque variables were significantly related to sprint performance. Notably, it has been suggested that both early (<100 ms) and late (> 100 ms) time points are governed by distinct physiological characteristics, such as neural activation and MVC force, respectively (7,14,15,80). The relationship between later time points and sprint performance from the current study implies that maximal strength may be a significant determinant of sprint performance, which is further evidenced by the significant relationship between MVC torque and sprint time from this study (r = -0.593, $P \leq 0.001$). In agreement with our findings, a study assessing isometric mid-thigh pull force found a significant relationship between 10 m sprint time with peak force (r = -0.673, $P \leq 0.01$) and rate of force development from 0-200 seconds (r = -0.451, $P \leq 0.05$) (102). A potential explanation for the negative relationship between maximal strength and sprint time is that greater ground reaction forces rather than speed of limb repositioning has been shown to be more indicative of faster running speed (103). However, several previous studies have assessed the relationship between rapid force variables and sprint performance and reported contrasting results (4,104). For example, lower sprint time during both five m and 20 m sprints was related to higher force at 100 ms (five m: r = -0.50, $P = 0.034$; 20 m: r = -0.50, $P = 0.037$) during an isometric squat, but neither were related to force at 50 ms, 150 ms, 200 ms, or max (4). Similarly, higher force at 100 ms during an isometric mid-thigh pull was related to lower 10 m sprint time (r= -0.54, $P < 0.01$), but not peak force (104). An important distinction between the studies that found a significant relationship between sprint

time and peak strength is that they included both males and females, while both studies that showed a relationship between force at 100 ms and sprint time only included males. Several sex differences exist during the early part of a sprint, including males having a shorter contact time for the first two steps (105). This shorter ground contact time may make them less reliant on maximal force for sprint performance and thus may explain the difference with the current study.

The lack of a relationship between the LE_{RSF} values and sprint time can likely be attributed to controlling for critical elements responsible for differentiating those better equipped to perform a sprint. For example, the specific force measurement quantifies the amount of force a muscle can produce per unit area (22,26), therefore eliminating the influence of PCSA, which has been shown to be a primary determinate of muscle strength (106). As previously stated, higher maximal strength in the current study was negatively related to lower sprint time and thus controlling for PCSA may limit the influence of this relationship. Further, longer FL has also been reported to be related to faster 100 m sprint time (107). While this sprint distance is much longer than used in the current study, it highlights the potential importance of factors extrinsic to the muscle itself on determining sprint performance. Lastly, the lack of relationship between body mass normalized rapid force and sprint time is supported by a previous study that did not see a change in the relationships between absolute and normalized rapid force from isometric squats and sprint time (4). However, these results are in contrast to previous studies that have examined improvements in relationships between rapid force variables and functional tasks when normalize to body mass (11,104).

In summary, eight regression equations were created to assess FL and PA shortly after the onset of contraction of each of the four constituent muscles of the quadriceps femoris, and utilized to create a model to assess rapid specific force of the quadriceps femoris muscle in

young healthy adults. The reliability of this novel measure increased with time from contraction onset, which followed the same general trend as the reliability of rapid absolute torque variables. It is likely that the overall lower reliability of the LE_{RSF} measure when compared to maximal strength measurements is due to both the lower overall reliability of specific force measurements (27) and of rapid strength measurements (4,100,101). Further, only absolute rapid torque at late time points were related to sprint time, potentially due to the relationship between maximal strength and sprint performance. Thus, the LE_{RSF} calculation, which provides a measure of intrinsic force producing capacity of the muscle, may control for several critical factors extrinsic to the muscle that determine sprint performance. Future studies are needed to determine if the LE_{RSF} measurement is predictive of other performance metrics or if it can account for sex-based differences in rapid strength (108).

Figure 1: A schematic of the experimental design. Visits two and three were identical, with visit two including additional torque tracings to determine the muscle architecture regression equations.

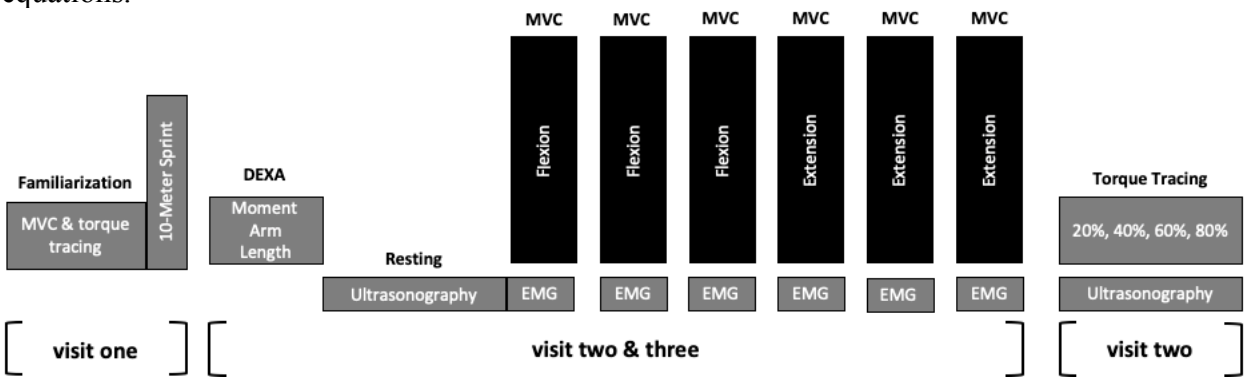

Figure 2: Graphical representations of the regression equations used to calculate fascicle length of the A) vastus lateralis, B) vastus intermedius, C) rectus femoris, and D) vastus medialis; MVC, maximal voluntary contraction.

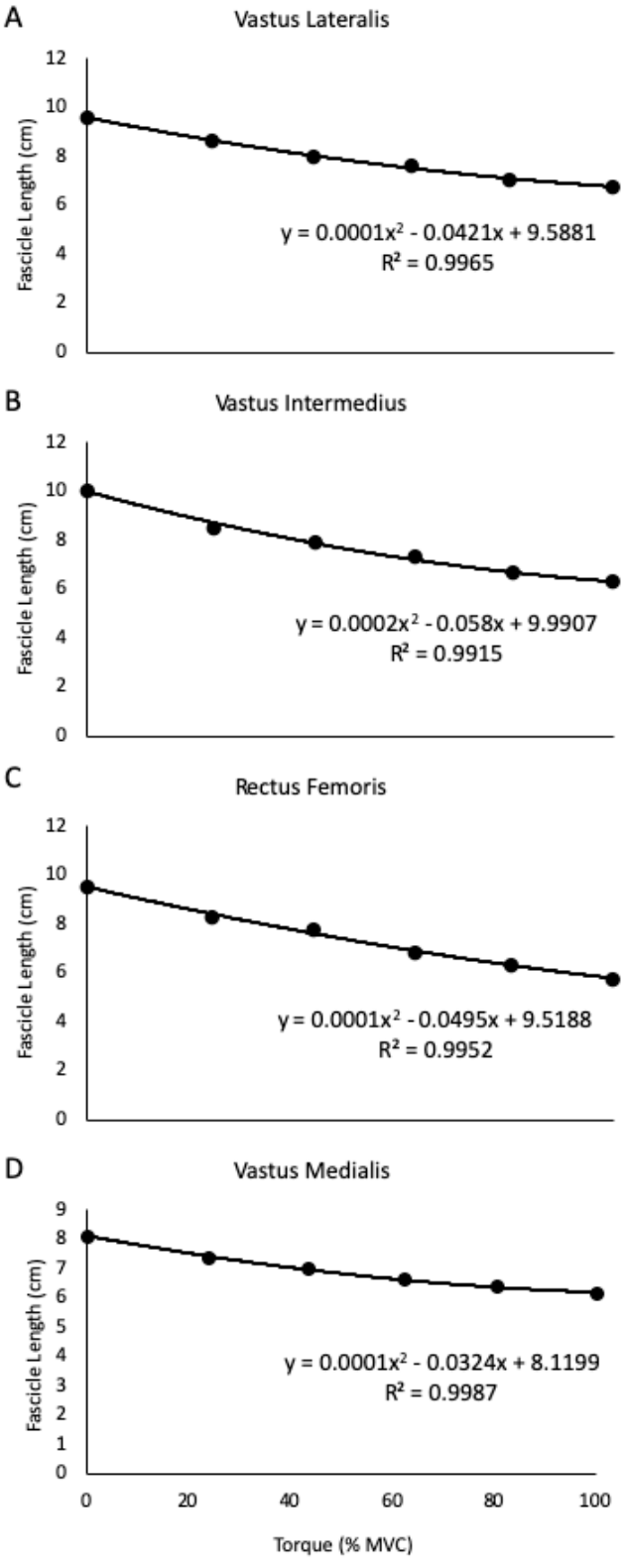

Figure 3: Graphical representations of the regression equations used to calculate pennation angle of the A) vastus lateralis, B) vastus intermedius, C) rectus femoris, and D) vastus medialis; MVC, maximal voluntary contraction.

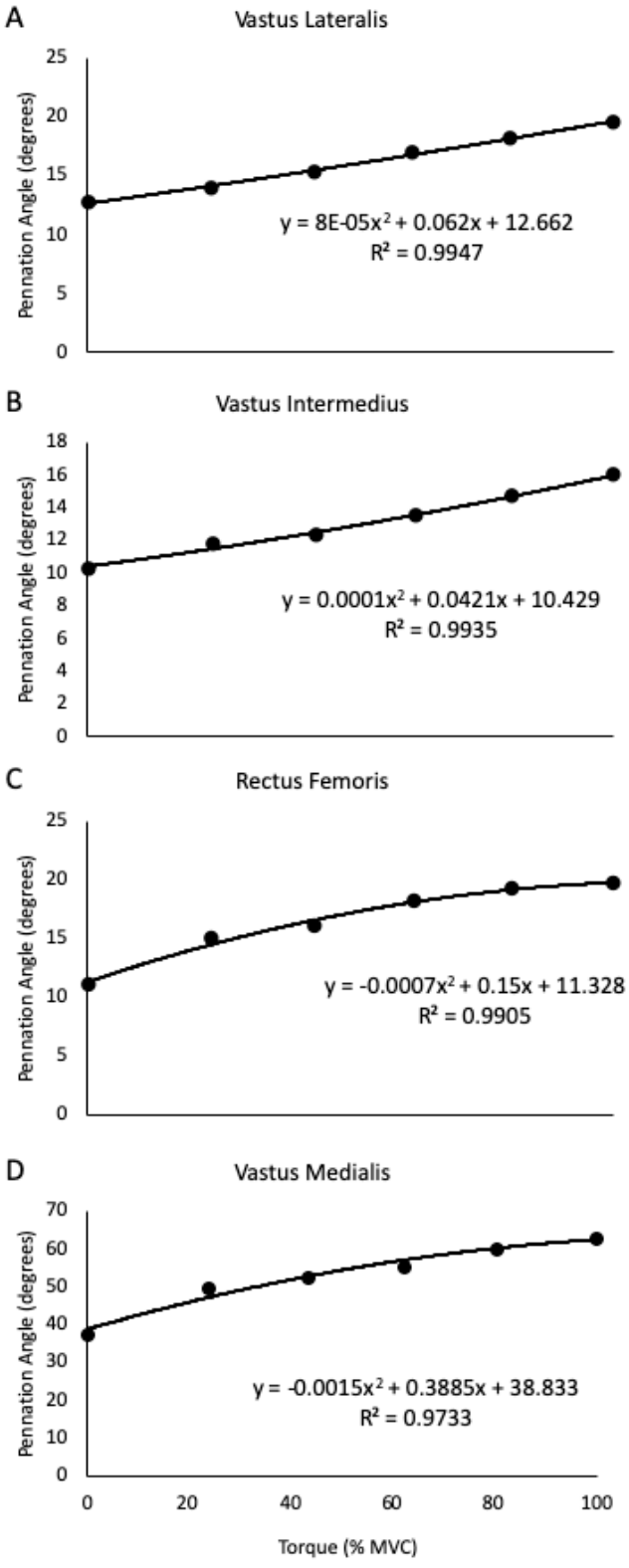

Figure 4: Graphical representations of the relationship between sprint time and absolute rapid torque at A) 150 ms and B) 200 ms.

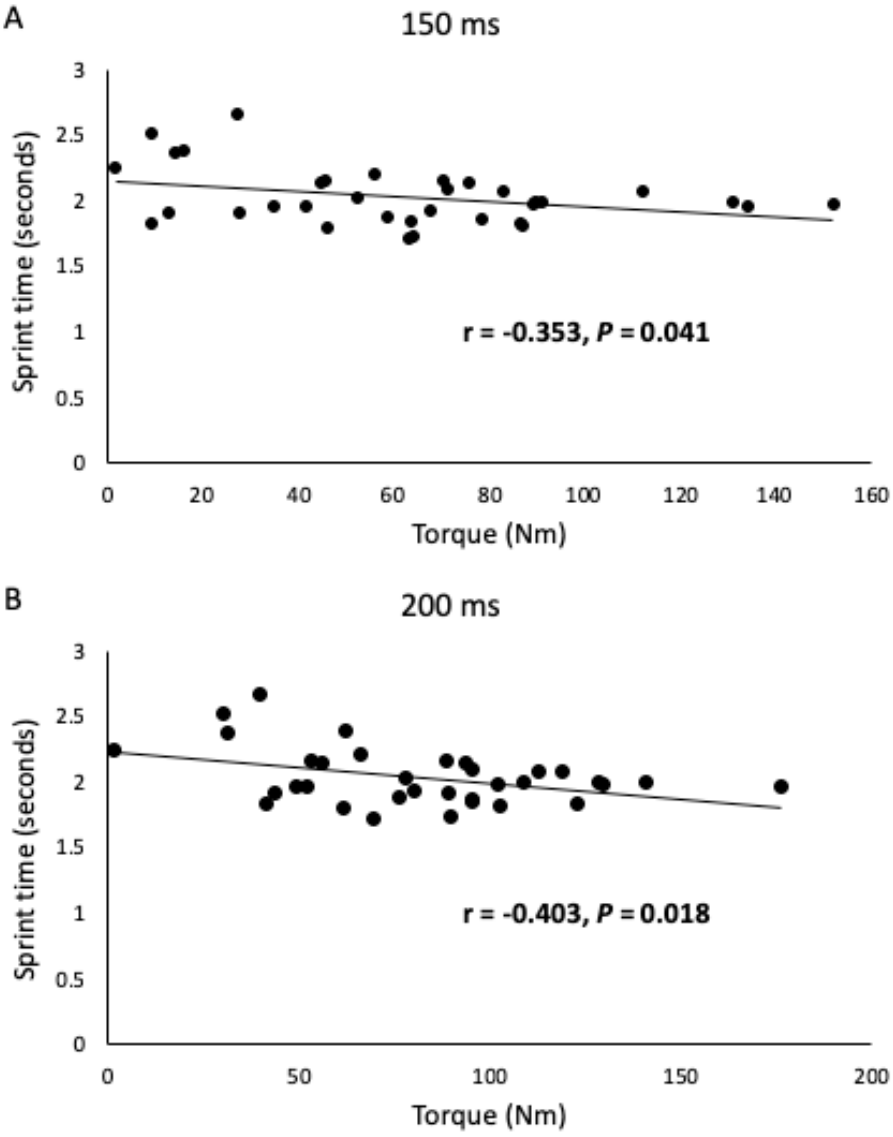

Table 1: Leg extensor rapid specific force and absolute torque values at each rapid time point from testing days two and three. All data presented as mean ± standard deviation. LE_{RSF}: Leg extensor rapid specific force, V2: visit 2, V3: visit 3. n = 33

	LE_{RSF} ($N \cdot cm^{-2}$)		Absolute rapid torque (Nm)	
	V2	V3	V2	V3
50 ms	1.03 ± 0.73	1.13 ± 1.06	4.75 ± 3.08	5.20 ± 3.05
100 ms	5.94 ± 4.89	5.98 ± 4.02	37.60 ± 35.72	39.58 ± 28.37
150 ms	9.48 ± 5.36	9.13 ± 4.54	61.09 ± 39.23	61.66 ± 36.57
200 ms	12.08 ± 5.24	11.35 ± 4.25	81.12 ± 39.87	79.93 ± 40.60

Table 2: Leg extensor rapid specific force, absolute torque, and normalized values at each rapid time point from testing day two used to assess the relationship with sprint time. All data presented as mean ± standard deviation. LE_{RSF}: Leg extensor rapid specific force. n = 34

	LE_{RSF} (N • cm^{-2})	Absolute rapid torque (Nm)	LE_{RSF} normalized (N • cm^{-2} / kg)	Absolute rapid torque normalized (Nm • kg^{-1})
50 ms	1.05 ± 0.74	4.71 ± 3.06	0.02 ± 0.01	0.07 ± 0.05
100 ms	6.01 ± 4.84	37.38 ± 35.21	0.09 ± 0.08	0.53 ± 0.47
150 ms	9.91 ± 5.35	61.90 ± 37.22	0.15 ± 0.09	0.89 ± 0.51
200 ms	12.60 ± 5.06	82.16 ± 36.71	0.19 ± 0.09	1.17 ± 0.50

REFERENCES

1. Maughan RJ, Watson JS, Weir J. Strength and cross-sectional area of human skeletal muscle. *J Physiol*. 1983;338(1):37-49. doi:10.1113/jphysiol.1983.sp014658

2. Clark DJ, Manini TM, Fielding RA, Patten C. Neuromuscular determinants of maximum walking speed in well-functioning older adults. *Exp Gerontol*. 2013;48(3):358-363. doi:10.1016/j.exger.2013.01.010

3. Hannah R, Minshull C, Smith SL, Folland JP. Longer electromechanical delay impairs hamstrings explosive force versus quadriceps. *Med Sci Sports Exerc*. 2014;46(5):963-972. doi:10.1249/MSS.0000000000000188

4. Tillin NA, Pain MTG, Folland J. Explosive force production during isometric squats correlates with athletic performance in rugby union players. *J Sports Sci*. 2013;31(1):66-76. doi:10.1080/02640414.2012.720704

5. Zebis MK, Andersen LL, Ellingsgaard H, Aagaard P. Rapid hamstring/quadriceps force capacity in male vs. female elite soccer players. *J Strength Cond Res*. 2011;25(7):1989-1993. doi:10.1519/JSC.0b013e3181e501a6

6. Maffiuletti NA, Bizzini M, Widler K, Munzinger U. Asymmetry in quadriceps rate of force development as a functional outcome measure in TKA. *Clin Orthop Relat Res*. 2010;468(1):191-198. doi:10.1007/s11999-009-0978-4

7. Maffiuletti NA, Aagaard P, Blazevich AJ, Folland J, Tillin N, Duchateau J. Rate of force development: physiological and methodological considerations. *Eur J Appl Physiol*. 2016;116(6):1091-1116. doi:10.1007/s00421-016-3346-6

8. Mota JA, Barnette TJ, Gerstner GR, et al. Relationships Between Neuromuscular Function and Functional Balance Performance in Firefighters. *Sci Rep*. 2018;8(1). doi:10.1038/s41598-018-33555-z

9. Palmer TB, Thiele RM, Williams KB, et al. The identification of fall history using maximal and rapid isometric torque characteristics of the hip extensors in healthy, recreationally active elderly females: A preliminary investigation. *Aging Clin Exp Res*. 2015;27(4):431-438. doi:10.1007/s40520-014-0305-0

10. Palmer TB, Hawkey MJ, Thiele RM, et al. The influence of athletic status on maximal and rapid isometric torque characteristics and postural balance performance in Division I female soccer athletes and non-athlete controls. *Clin Physiol Funct Imaging*. 2015;35(4):314-322. doi:10.1111/cpf.12167

11. Thompson BJ, Ryan ED, Sobolewski EJ, et al. Relationships between rapid isometric torque characteristics and vertical jump performance in division i collegiate American football players: Influence of body mass normalization. *J Strength Cond Res*.

2013;27(10):2737-2742. doi:10.1519/JSC.0b013e318281637b

12. Thompson BJ, Ryan ED, Sobolewski EJ, et al. Can maximal and rapid isometric torque characteristics predict playing level in division i american collegiate football players? *J Strength Cond Res*. 2013;27(3):655-661. doi:10.1519/JSC.0b013e31825bb56c

13. Tillin NA, Jimenez-Reyes P, Pain MTG, Folland JP. Neuromuscular performance of explosive power athletes versus untrained individuals. *Med Sci Sports Exerc*. 2010;42(4):781-790. doi:10.1249/MSS.0b013e3181be9c7e

14. Aagaard P, Simonsen EB, Andersen JL, Magnusson P, Dyhre-Poulsen P. Increased rate of force development and neural drive of human skeletal muscle following resistance training. *J Appl Physiol*. 2002;93(4):1318-1326. doi:10.1152/japplphysiol.00283.2002

15. Andersen LL, Aagaard P. Influence of maximal muscle strength and intrinsic muscle contractile properties on contractile rate of force development. *Eur J Appl Physiol*. 2006;96(1):46-52. doi:10.1007/s00421-005-0070-z

16. Tsaopoulos DE, Baltzopoulos V, Maganaris CN. Human patellar tendon moment arm length: Measurement considerations and clinical implications for joint loading assessment. *Clin Biomech*. 2006;21(7):657-667. doi:10.1016/j.clinbiomech.2006.02.009

17. Kellis E, Baltzopoulos V. The effects of antagonist moment on the resultant knee joint moment during isokinetic testing of the knee extensors. *Eur J Appl Physiol Occup Physiol*. 1997;76(3):253-259. doi:10.1007/s004210050244

18. Massey G, Evangelidis P, Folland J. Influence of contractile force on the architecture and morphology of the quadriceps femoris. *Exp Physiol*. 2015;100(11):1342-1351. doi:10.1113/EP085360

19. Stebbings GK, Morse CI, Williams AG, Day SH. Variability and distribution of muscle strength and its determinants in humans. *Muscle and Nerve*. 2014;49(6):879-886. doi:10.1002/mus.24075

20. Erskine RM, Jones DA, Maffulli N, Williams AG, Stewart CE, Degens H. What causes in vivo muscle specific tension to increase following resistance training? *Exp Physiol*. 2011;96(2):145-155. doi:10.1113/expphysiol.2010.053975

21. Morse CI, Thom JM, Reeves ND, Birch KM, Narici M V. In vivo physiological cross-sectional area and specific force are reduced in the gastrocnemius of elderly men. *J Appl Physiol*. 2005;99(3):1050-1055. doi:10.1152/japplphysiol.01186.2004

22. Reeves ND, Narici M V., Maganaris CN. Effect of resistance training on skeletal muscle-specific force in elderly humans. *J Appl Physiol*. 2004;96(3):885-892. doi:10.1152/japplphysiol.00688.2003

23. Erskine RM, Jones DA, Williams AG, Stewart CE, Degens H. Resistance training increases in vivo quadriceps femoris muscle specific tension in young men. *Acta Physiol*. 2010;199(1):83-89. doi:10.1111/j.1748-1716.2010.02085.x

24. De Ruiter CJ, Vermeulen G, Toussaint HM, De Haan A. Isometric knee-extensor torque development and jump height in volleyball players. *Med Sci Sports Exerc*. 2007;39(8):1336-1346. doi:10.1097/mss.0b013e318063c719

25. Van Der Zwaard S, Van Der Laarse WJ, Weide G, et al. Critical determinants of combined sprint and endurance performance: An integrative analysis from muscle fiber to the human body. *FASEB J*. 2018;32(4):2110-2123. doi:10.1096/fj.201700827R

26. Maganaris CN, Baltzopoulos V, Ball D, Sargeant AJ. In vivo specific tension of human skeletal muscle. *J Appl Physiol*. 2001;90(3):865-872. doi:10.1152/jappl.2001.90.3.865

27. Erskine RM, Jones DA, Maganaris CN, Degens H. In vivo specific tension of the human quadriceps femoris muscle. *Eur J Appl Physiol*. 2009;106(6):827-838. doi:10.1007/s00421-009-1085-7

28. Erskine RM, Jones DA, Williams AG, Stewart CE, Degens H. Inter-individual variability in the adaptation of human muscle specific tension to progressive resistance training. *Eur J Appl Physiol*. 2010;110(6):1117-1125. doi:10.1007/s00421-010-1601-9

29. Morse CI, Tolfrey K, Thom JM, Vassilopoulos V, Maganaris CN, Narici M V. Gastrocnemius muscle specific force in boys and men. *J Appl Physiol*. 2008;104(2):469-474. doi:10.1152/japplphysiol.00697.2007

30. Sims DT, Onambélé-Pearson GL, Burden A, Payton C, Morse CI. Specific force of the vastus lateralis in adults with achondroplasia. *J Appl Physiol*. 2018;124(3):696-703. doi:10.1152/japplphysiol.00638.2017

31. Widrick JJ, Stelzer JE, Shoepe TC, Garner DP. Functional properties of human muscle fibers after short-term resistance exercise training. *Am J Physiol - Regul Integr Comp Physiol*. 2002;283(2 52-2). doi:10.1152/ajpregu.00120.2002

32. Luden N, Minchev K, Hayes E, Louis E, Trappe T, Trappe S. Human vastus lateralis and soleus muscles display divergent cellular contractile properties. *Am J Physiol - Regul Integr Comp Physiol*. 2008;295(5):R1593. doi:10.1152/ajpregu.90564.2008

33. Parente V, D'Antona G, Adami R, et al. Long-term resistance training improves force and unloaded shortening velocity of single muscle fibres of elderly women. *Eur J Appl Physiol*. 2008;104(5):885-893. doi:10.1007/s00421-008-0845-0

34. Bottinelli R, Canepari M, Pellegrino MA, Reggiani C. Force-velocity properties of human skeletal muscle fibres: Myosin heavy chain isoform and temperature dependence. *J Physiol*. 1996;495(2):573-586. doi:10.1113/jphysiol.1996.sp021617

35. D'Antona G, Lanfranconi F, Pellegrino MA, et al. Skeletal muscle hypertrophy and structure and function of skeletal muscle fibres in male body builders. *J Physiol*. 2006;570(3):611-627. doi:10.1113/jphysiol.2005.101642

36. Pansarasa O, Rinaldi C, Parente V, Miotti D, Capodaglio P, Bottinelli R. Resistance training of long duration modulates force and unloaded shortening velocity of single muscle fibres of young women. *J Electromyogr Kinesiol*. 2009;19(5):e290-e300. doi:10.1016/j.jelekin.2008.07.007

37. Tikunov BA, Sweeney HL, Rome LC. Quantitative electrophoretic analysis of myosin heavy chains in single muscle fibers. *J Appl Physiol*. 2001;90(5):1927-1935. doi:10.1152/jappl.2001.90.5.1927

38. Reeves ND, Narici M V., Maganaris CN. Effect of resistance training on skeletal muscle-specific force in elderly humans. *J Appl Physiol*. 2004;96(3):885-892. doi:10.1152/japplphysiol.00688.2003

39. O'Brien TD, Reeves ND, Baltzopoulos V, Jones DA, Maganaris CN. In vivo measurements of muscle specific tension in adults and children. *Exp Physiol*. 2010;95(1):202-210. doi:10.1113/expphysiol.2009.048967

40. Narici M V., Landoni L, Minetti AE. Assessment of human knee extensor muscles stress from in vivo physiological cross-sectional area and strength measurements. *Eur J Appl Physiol Occup Physiol*. 1992;65(5):438-444. doi:10.1007/BF00243511

41. Pandy MG. Moment Arm of a Muscle Force. *Exerc Sport Sci Rev*. 1999;27(1):79-118.

42. Tsaopoulos DE, Baltzopoulos V, Richards PJ, Maganaris CN. In vivo changes in the human patellar tendon moment arm length with different modes and intensities of muscle contraction. *J Biomech*. 2007;40(15):3325-3332. doi:10.1016/j.jbiomech.2007.05.005

43. Enoka RM, Duchateau J. Rate coding and the control of muscle force. *Cold Spring Harb Perspect Med*. 2017;7(10). doi:10.1101/cshperspect.a029702

44. Van Cutsem M, Duchateau J. Preceding muscle activity influences motor unit discharge and rate of torque development during ballistic contractions in humans. *J Physiol*. 2005;562(2):635-644. doi:10.1113/jphysiol.2004.074567

45. Oya T, Riek S, Cresswell AG. Recruitment and rate coding organisation for soleus motor units across entire range of voluntary isometric plantar flexions. *J Physiol*. 2009;587(19):4737-4748. doi:10.1113/jphysiol.2009.175695

46. Van Cutsem M, Duchateau J, Hainaut K. Changes in single motor unit behaviour contribute to the increase in contraction speed after dynamic training in humans. *J Physiol*. 1998;513(1):295-305. doi:10.1111/j.1469-7793.1998.295by.x

47. Klass M, Baudry S, Duchateau J. Age-related decline in rate of torque development is accompanied by lower maximal motor unit discharge frequency during fast contractions. *J Appl Physiol*. 2008;104(3):739-746. doi:10.1152/japplphysiol.00550.2007

48. Grimby L, Hannerz J, Hedman B. The fatigue and voluntary discharge properties of single motor units in man. *J Physiol*. 1981;316(1):545-554. doi:10.1113/jphysiol.1981.sp013805

49. Shield A, Zhou S. Assessing voluntary muscle activation with the twitch interpolation technique. *Sport Med*. 2004;34(4):253-267. doi:10.2165/00007256-200434040-00005

50. Millet GY, Martin V, Martin A, Vergès S. Electrical stimulation for testing neuromuscular function: From sport to pathology. *Eur J Appl Physiol*. 2011;111(10):2489-2500. doi:10.1007/s00421-011-1996-y

51. Rutherford OM, Jones DA, Newham DJ. Clinical and experimental application of the percutaneous twitch superimposition technique for the study of human muscle activation. *J Neurol Neurosurg Psychiatry*. 1986;49(11):1288-1291. doi:10.1136/jnnp.49.11.1288

52. Mau-Moeller A, Behrens M, Lindner T, Bader R, Bruhn S. Age-related changes in neuromuscular function of the quadriceps muscle in physically active adults. *J Electromyogr Kinesiol*. 2013;23(3):640-648. doi:10.1016/j.jelekin.2013.01.009

53. Roos MR, Rice CL, Connelly DM, Vandervoort AA. Quadriceps muscle strength, contractile properties, and motor unit firing rates in young and old men. *Muscle and Nerve*. 1999;22(8):1094-1103. doi:10.1002/(SICI)1097-4598(199908)22:8<1094::AID-MUS14>3.0.CO;2-G

54. Rozand V, Sundberg CW, Hunter SK, Smith AE. Age-related Deficits in Voluntary Activation: A Systematic Review and Meta-analysis. *Med Sci Sports Exerc*. 2020;52(3):549-560. doi:10.1249/MSS.0000000000002179

55. Baratta R, Solomonow M, Zhou BH, Letson D, Chuinard R, D'Ambrosia R. Muscular coactivation. The role of the antagonist musculature in maintaining knee stability. *Am J Sports Med*. 1988;16(2):113-122. doi:10.1177/036354658801600205

56. Izquierdo M, Ibañez J, Gorostiaga E, et al. Maximal strength and power characteristics in isometric and dynamic actions of the upper and lower extremities in middle-aged and older men. *Acta Physiol Scand*. 1999;167(1):57-68. doi:10.1046/j.1365-201x.1999.00590.x

57. Macaluso A, Nimmo MA, Foster JE, Cockburn M, McMillan NC, De Vito G. Contractile muscle volume and agonist-antagonist coactivation account for differences in torque between young and older women. *Muscle and Nerve*. 2002;25(6):858-863. doi:10.1002/mus.10113

58. Morse CI, Thom JM, Mian OS, Muirhead A, Birch KM, Narici M V. Muscle strength,

volume and activation following 12-month resistance training in 70-year-old males. *Eur J Appl Physiol*. 2005;95(2-3):197-204. doi:10.1007/s00421-005-1342-3

59. Morse CI, Thom JM, Mian OS, Birch KM, Narici M V. Gastrocnemius specific force is increased in elderly males following a 12-month physical training programme. *Eur J Appl Physiol*. 2007;100(5):563-570. doi:10.1007/s00421-006-0246-1

60. Häkkinen K, Kallinen M, Izquierdo M, et al. Changes in agonist-antagonist EMG, muscle CSA, and force during strength training in middle-aged and older people. *J Appl Physiol*. 1998;84(4):1341-1349. doi:10.1152/jappl.1998.84.4.1341

61. De Boer MD, Morse CI, Thom JM, De Haan A, Narici M V. Changes in antagonist muscles' coactivation in response to strength training in older women. *Journals Gerontol - Ser A Biol Sci Med Sci*. 2007;62(9):1022-1027. doi:10.1093/gerona/62.9.1022

62. Powell PL, Roy RR, Kanim P, Bello MA, Edgerton VR. Predictability of skeletal muscle tension from architectural determinations in guinea pig hindlimbs. *J Appl Physiol Respir Environ Exerc Physiol*. 1984;57(6):1715-1721. doi:10.1152/jappl.1984.57.6.1715

63. Trezise J, Collier N, Blazevich AJ. Anatomical and neuromuscular variables strongly predict maximum knee extension torque in healthy men. *Eur J Appl Physiol*. 2016;116(6):1159-1177. doi:10.1007/s00421-016-3352-8

64. Kawakami Y, Abe T, Fukunaga T. Muscle-fiber pennation angles are greater in hypertrophied than in normal muscles. *J Appl Physiol*. 1993;74(6):2740-2744. doi:10.1152/jappl.1993.74.6.2740

65. Sacks RD, Roy RR. Architecture of the hind limb muscles of cats: Functional significance. *J Morphol*. 1982;173(2):185-195. doi:10.1002/jmor.1051730206

66. Narici M V., Maganaris CN, Reeves ND, Capodaglio P. Effect of aging on human muscle architecture. *J Appl Physiol*. 2003;95(6):2229-2234. doi:10.1152/japplphysiol.00433.2003

67. Karamanidis K, Arampatzis A. Mechanical and morphological properties of human quadriceps femoris and triceps surae muscle-tendon unit in relation to aging and running. *J Biomech*. 2006;39(3):406-417. doi:10.1016/j.jbiomech.2004.12.017

68. Wickiewicz TL, Roy RR, Powell PL, Edgerton VR. Muscle architecture of the human lower limb. *Clin Orthop Relat Res*. 1983;No. 179(179):275-283. doi:10.1097/00003086-198310000-00042

69. Haxton HA. Absolute muscle force in the ankle flexors of man. *J Physiol*. 1944;103(3):267-273. doi:10.1113/jphysiol.1944.sp004075

70. Friederich JA, Brand RA. Muscle fiber architecture in the human lower limb. *J Biomech*. 1990;23(1):91-95. doi:10.1016/0021-9290(90)90373-B

71. Hogrel JY, Barnouin Y, Azzabou N, et al. NMR imaging estimates of muscle volume and intramuscular fat infiltration in the thigh: variations with muscle, gender, and age. *Age (Omaha)*. 2015;37(3):1-11. doi:10.1007/s11357-015-9798-5

72. Morse CI, Degens H, Jones DA. The validity of estimating quadriceps volume from single MRI cross-sections in young men. *Eur J Appl Physiol*. 2007;100(3):267-274. doi:10.1007/s00421-007-0429-4

73. Lund H, Christensen L, Savnik A, Boesen J, Danneskiold-Samsøe B, Bliddal H. Volume estimation of extensor muscles of the lower leg based on MR imaging. *Eur Radiol*. 2002;12(12):2982-2987. doi:10.1007/s00330-002-1334-1

74. Körting C, Schlippe M, Petersson S, et al. In vivo muscle morphology comparison in post-stroke survivors using ultrasonography and diffusion tensor imaging. *Sci Rep*. 2019;9(1). doi:10.1038/s41598-019-47968-x

75. Huxley H, Hanson J. Changes in the Cross-striations of muscle during contraction and stretch and their structural interpretation. *Nature*. 1954;173(4412):973-976. doi:10.1038/173973a0

76. Robert S, Herman JR, Schuenke MD. Misclassification of hybrid fast fibers in resistance-trained human skeletal muscle using histochemical and immunohistochemical methods. *J Strength Cond Res*. 2012;26(10):2616-2622. doi:10.1519/JSC.0b013e3182667095

77. Kelly NA, Hammond KG, Stec MJ, et al. Quantification and characterization of grouped type I myofibers in human aging. *Muscle and Nerve*. 2018;57(1):E52-E59. doi:10.1002/mus.25711

78. Sonjak V, Jacob K, Morais JA, et al. Fidelity of muscle fibre reinnervation modulates aging muscle impact in elderly women. *J Physiol*. August 2019:JP278261. doi:10.1113/JP278261

79. Gerstner GR, Thompson BJ, Rosenberg JG, Sobolewski EJ, Scharville MJ, Ryan ED. Neural and Muscular Contributions to the Age-Related Reductions in Rapid Strength. *Med Sci Sports Exerc*. 2017;49(7):1331-1339. doi:10.1249/MSS.0000000000001231

80. Folland JP, Buckthorpe MW, Hannah R. Human capacity for explosive force production: Neural and contractile determinants. *Scand J Med Sci Sport*. 2014;24(6):894-906. doi:10.1111/sms.12131

81. De Ruiter CJ, Kooistra RD, Paalman MI, De Haan A. Initial phase of maximal voluntary and electrically stimulated knee extension torque development at different knee angles. *J Appl Physiol*. 2004;97(5):1693-1701. doi:10.1152/japplphysiol.00230.2004

82. Tillin NA, Pain MTG, Folland JP. Contraction type influences the human ability to use the available torque capacity of skeletal muscle during explosive efforts. *Proc R Soc B Biol*

Sci. 2012;279(1736):2106-2115. doi:10.1098/rspb.2011.2109

83. Kulig K, Andrews JG, Hay JG. Human strength curves. *Exerc Sport Sci Rev*. 1984;12(1):417-466. doi:10.1249/00003677-198401000-00014

84. Harridge SDR, Bottinelli R, Canepari M, et al. Whole-muscle and single-fibre contractile properties and myosin heavy chain isoforms in humans. *Pflugers Arch Eur J Physiol*. 1996;432(5):913-920. doi:10.1007/s004240050215

85. Duchateau J, Baudry S. Maximal discharge rate of motor units determines the maximal rate of force development during ballistic contractions in human. *Front Hum Neurosci*. 2014;8(1 APR). doi:10.3389/fnhum.2014.00234

86. Bojsen-Møller J, Magnusson SP, Rasmussen LR, Kjaer M, Aagaard P. Muscle performance during maximal isometric and dynamic contractions is influenced by the stiffness of the tendinous structures. *J Appl Physiol*. 2005;99(3):986-994. doi:10.1152/japplphysiol.01305.2004

87. Tillin NA, Folland JP. Maximal and explosive strength training elicit distinct neuromuscular adaptations, specific to the training stimulus. *Eur J Appl Physiol*. 2014;114(2):365-374. doi:10.1007/s00421-013-2781-x

88. Thompson BJ, Ryan ED, Sobolewski EJ, Conchola EC, Cramer JT. Age related differences in maximal and rapid torque characteristics of the leg extensors and flexors in young, middle-aged and old men. *Exp Gerontol*. 2013;48(2):277-282. doi:10.1016/j.exger.2012.10.009

89. Weir JP. Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. *J Strength Cond Res*. 2005;19(1):231-240. doi:10.1519/15184.1

90. Hopkins WG. Measures of reliability in sports medicine and science. *Sport Med*. 2000;30(1):1-15. doi:10.2165/00007256-200030010-00001

91. Shrout PE, Fleiss JL. Intraclass correlations: Uses in assessing rater reliability. *Psychol Bull*. 1979;86(2):420-428. doi:10.1037/0033-2909.86.2.420

92. Thompson BJ. Influence of signal filtering and sample rate on isometric torque – time parameters using a traditional isokinetic dynamometer. *J Biomech*. 2019;83:235-242. doi:10.1016/j.jbiomech.2018.12.006

93. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol*. 2000;10(5):361-374. doi:10.1016/S1050-6411(00)00027-4

94. Ryan ED, Rosenberg JG, Scharville MJ, Sobolewski EJ, Tweedell AJ, Kleinberg CR. Pennation Angle Does Not Influence the Age-Related Differences in Echo Intensity of the

Medial Gastrocnemius. *Ultrasound Med Biol*. 2015;41(2):618-621. doi:10.1016/j.ultrasmedbio.2014.08.004

95. Massey G, Evangelidis P, Folland J. Influence of contractile force on the architecture and morphology of the quadriceps femoris. *Exp Physiol*. 2015;100(11):1342-1351. doi:10.1113/EP085360

96. Maganaris CN, Baltzopoulos V, Ball D, Sargeant AJ. In vivo specific tension of human skeletal muscle. *J Appl Physiol*. 2001;90(3):865-872. doi:10.1152/jappl.2001.90.3.865

97. Steiger JH. Tests for comparing elements of a correlation matrix. *Psychol Bull*. 1980;87(2):245-251. doi:10.1037/0033-2909.87.2.245

98. Noorkoiv M, Stavnsbo A, Aagaard P, Blazevich AJ. In vivo assessment of muscle fascicle length by extended field-of-view ultrasonography. *J Appl Physiol*. 2010;109(6):1974-1979. doi:10.1152/japplphysiol.00657.2010

99. Gallina A, Render JN, Santos J, et al. Influence of knee joint position and sex on vastus medialis regional architecture. *Appl Physiol Nutr Metab*. 2018;43(6):643-646. doi:10.1139/apnm-2017-0697

100. Buckthorpe MW, Hannah R, Pain TG, Folland JP. Reliability of neuromuscular measurements during explosive isometric contractions, with special reference to electromyography normalization techniques. *Muscle and Nerve*. 2012;46(4):566-576. doi:10.1002/mus.23322

101. Thompson BJ, Ryan ED, Herda TJ, et al. Consistency of rapid muscle force characteristics: Influence of muscle contraction onset detection methodology. *J Electromyogr Kinesiol*. 2012;22(6):893-900. doi:10.1016/j.jelekin.2012.05.008

102. Townsend JR, Bender D, Vantrease WC, et al. Isometric Midthigh Pull Performance Is Associated With Athletic Performance and Sprinting Kinetics in Division I Men and Women's Basketball Players. *J strength Cond Res*. 2019;33(10):2665-2673. doi:10.1519/JSC.0000000000002165

103. Weyand PG, Sternlight DB, Bellizzi MJ, Wright S. Faster top running speeds are achieved with greater ground forces not more rapid leg movements. *J Appl Physiol*. 2000;89(5):1991-1999. doi:10.1152/jappl.2000.89.5.1991

104. West DJ, Owen NJ, Jones MR, et al. Relationships between force-time characteristics of the isometric midthigh pull and dynamic performance in professional rugby league players. *J Strength Cond Res*. 2011;25(11):3070-3075. doi:10.1519/JSC.0b013e318212dcd5

105. Ciacci S, Merni F, Bartolomei S, Di Michele R. Sprint start kinematics during competition in elite and world-class male and female sprinters. *J Sports Sci*. 2017;35(13):1270-1278.

doi:10.1080/02640414.2016.1221519

106. Bamman MM, Newcomer BR, Larson-Meyer DE, Weinsier RL, Hunter GR. Evaluation of the strength-size relationship in vivo using various muscle size indices. *Med Sci Sports Exerc*. 2000;32(7):1307-1313. doi:10.1097/00005768-200007000-00019

107. Kumagai K, Abe T, Brechue WF, Ryushi T, Takano S, Mizuno M. Sprint performance is related to muscle fascicle length in male 100-m sprinters. *J Appl Physiol*. 2000;88(3):811-816. doi:10.1152/jappl.2000.88.3.811

108. Hannah R, Minshull C, Buckthorpe MW, Folland JP. Explosive neuromuscular performance of males versus females. *Exp Physiol*. 2012;97(5):618-629. doi:10.1113/expphysiol.2011.063420